

Closed-loop Control of Functional Neuromuscular Stimulation

NIH Neuroprosthesis Program Contract Number N01-NS-6-2338
Quarterly Progress Report #9
April 1, 1998 to June 30, 1998

Investigators:

Patrick E. Crago, Ph.D.
Clayton L. Van Doren, Ph.D.
Warren M. Grill, Ph.D.
Michael W. Keith, M.D.
Kevin L. Kilgore, Ph.D.
Joseph M. Mansour, Ph.D.
Wendy M. Murray, PhD.
P. Hunter Peckham, Ph.D.
David L. Wilson, Ph.D.

Departments of
Biomedical Engineering,
Mechanical and Aerospace Engineering,
and Orthopaedics
Case Western Reserve University
and MetroHealth Medical Center

**THIS QPR IS BEING SENT TO
YOU BEFORE IT HAS BEEN
REVIEWED BY THE STAFF OF THE
NEURAL PROSTHESIS PROGRAM.**

1. SYNTHESIS OF UPPER EXTREMITY FUNCTION.....	3
1. A. BIOMECHANICAL MODELING: PARAMETERIZATION AND VALIDATION.....	3
Purpose	3
Report of progress.....	3
1. a. i. <i>MOMENT ARMS VIA MAGNETIC RESONANCE IMAGING</i>	3
Abstract	3
Progress Report.....	3
Plans for next quarter.....	5
1.a.ii. <i>PASSIVE AND ACTIVE MOMENTS</i>	5
Abstract	5
Purpose	5
Report of Progress	5
Plans for Next Quarter.....	8
1. B. BIOMECHANICAL MODELING: ANALYSIS AND IMPROVEMENT OF GRASP OUTPUT.....	8
Abstract	8
Objective	8
2. CONTROL OF UPPER EXTREMITY FUNCTION.....	8
2. A. HOME EVALUATION OF CLOSED-LOOP CONTROL AND SENSORY FEEDBACK.....	8
Abstract	8
Purpose	8
Report of Progress	8
Plans for Next Quarter.....	11
2. B. INNOVATIVE METHODS OF CONTROL AND SENSORY FEEDBACK.....	14
2. b. i. <i>ASSESSMENT OF SENSORY FEEDBACK IN THE PRESENCE OF VISION</i>	14
Abstract	14
Purpose	14
Report of Progress	14
Plans for Next Quarter.....	15
2. b. ii. <i>INNOVATIVE METHODS OF COMMAND CONTROL</i>	15
Abstract	15
Purpose	15
Plans for Next Quarter.....	16
2. b. iii. <i>INCREASING WORKSPACE AND REPERTOIRE WITH BIMANUAL HAND GRASP</i>	16
Abstract	16
Purpose	16
Report of progress.....	16
Plans for Next Quarter.....	20
2. b. iv. <i>CONTROL OF HAND AND WRIST</i>	22
Abstract	22
Purpose	22
Report of progress.....	22
Plans for next quarter.....	23
References.....	23

1. SYNTHESIS OF UPPER EXTREMITY FUNCTION

The overall goals of this project are (1) to measure the biomechanical properties of the neuroprosthesis user's upper extremity and incorporate those measurements into a complete model with robust predictive capability, and (2) to use the predictions of the model to improve the grasp output of the hand neuroprosthesis for individual users.

1. a. BIOMECHANICAL MODELING: PARAMETERIZATION AND VALIDATION

Purpose

In this section of the contract, we will develop methods for obtaining biomechanical data from individual persons. Individualized data will form the basis for model-assisted implementation of upper extremity FNS. Using individualized biomechanical models, specific treatment procedures will be evaluated for individuals. The person-specific parameters of interest are tendon moment arms and lines of action, passive moments, and maximum active joint moments. Passive moments will be decomposed into components arising from stiffness inherent to a joint and from passive stretching of muscle-tendon units that cross one or more joints.

Report of progress

1. a. i. MOMENT ARMS VIA MAGNETIC RESONANCE IMAGING

Abstract

In this quarter, we completed a manuscript describing the estimation of tendon moment arms at the MP joint of the long finger. We have also begun to apply these techniques to analyze the biomechanics of the extensor carpi ulnaris to extensor carpi radialis brevis tendon transfer.

Progress Report

We hypothesize that after ECU to ECRB tendon transfer the extension moment vs. wrist angle will be the force generated by the ECU acting via the moment arm of the ECRB. The ECU length tension properties will be calculated from the moment-angle relationship measured by the Wrist Moment Transducer during stimulation (see Section 2.b.iv of this report), and the moment arm-angle relationship estimated from MRI images.

Estimation of the tendon moment arm using the tendon excursion method with 3D MRI images.

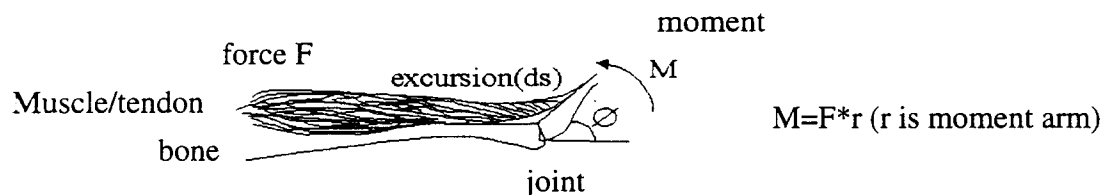


Fig. 1.a.i.1. Anatomical sketch of muscle crossing a single joint, defining the key variables in the kinematic and moment calculations.

The method of measuring moment arms from static MR images relies on the relationship between tendon excursion and joint angle (see Figure 1.a.i.1). When the joint rotates, the muscle elongates as the

tendon of insertion moves along with the distal anatomical segment. The larger the moment arm of the muscle at the joint, the larger the excursion of the tendon. The relation between excursion and angle can be derived from equating the work done by the muscle to the work done at the joint.

At a joint, the work done by the muscle tendon unit producing a force F to move the joint through an angle $d\Phi$ is given by $F \cdot ds = M \cdot d\Phi$, where ds is the tendon excursion and M is the joint torque. Since the moment is given by $M = F \cdot r$, the tendon moment arm r is $ds/d\Phi$. For the wrist which is a multi-segment structure, the moment arm r is an equivalent moment arm, and might not appear as the anatomical moment arm at any of the joints. However, the equivalent moment arm is what is desired in our calculations.

Thus, the moment arm can be approximated by the ratio of change in excursion to change in joint angle. If the path length is known at different joint angles, then the difference between path lengths is the excursion for that angle change. We have acquired one trial set of images at the wrist (see e.g. Figure 1.a.i.2). The wrist is imaged at neutral, flexion and extension angles. The ECU & ECRB tendons are visualized from these images, and the tendon paths are followed from slice to slice.

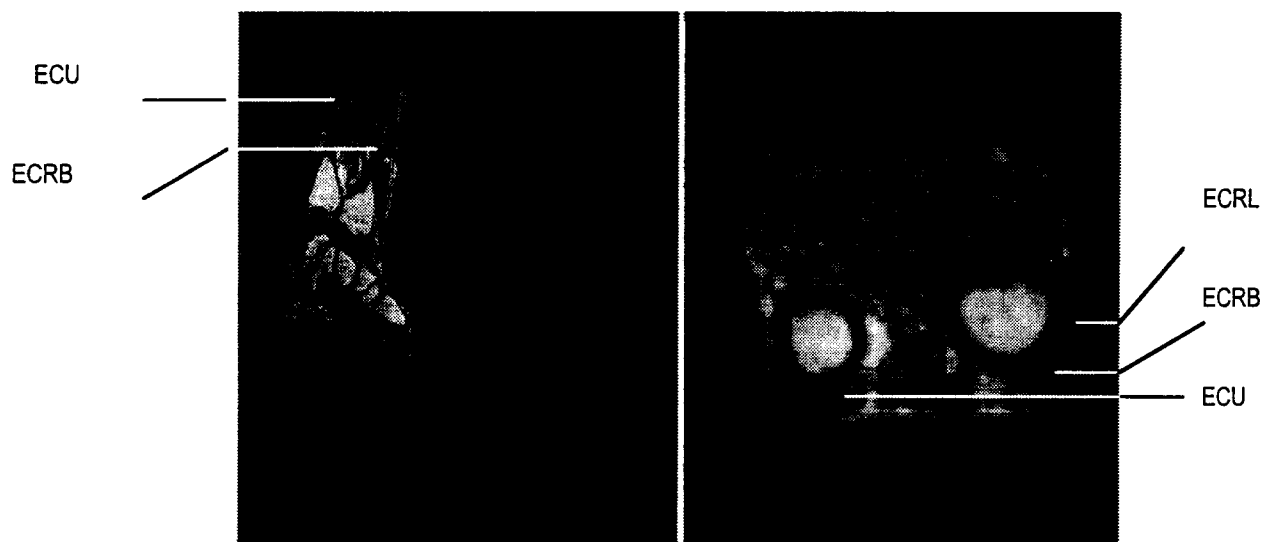


Figure 1.a.i.2. MR images of the wrist in extension. Left: coronal view; Right, transverse view proximal to the wrist.

With the wrist in flexion and extension the tendon at the distal end bends and glides along its path to the insertion point. To track the tendon along its curved path part we reformat the volume and obtain oblique views that enable us to follow the tendon. We can select the angle at which we want the oblique slices by input from the mouse. When using the standard transverse view and the oblique view there is chance of wrap around in the Z plane along the tendon which would give erroneous tendon lengths. We corrected for this by fixing the point about which we rotate so as not to overlap in the Z plane. One mouse button is used to follow the point in all 3 views and the other button is finally used to select the centroid of the tendon. Finally a least squares polynomial fit is used to interpolate between the centroid of each slice to calculate the tendon length at each position. The tendon excursion is taken as the difference in the tendon path lengths at successive joint angles.

The joint angle is obtained from segments of the bone. We ran several trials with various segments of the bone to compare measures. The method uses the primary axis of the bone along its length and

thus works best for long bones along relatively symmetric cylindrical parts. The best results are obtained when we use portions other than the head and styloid processes.

Plans for next quarter

We will continue the application of the biomechanical imaging analysis of tendon moment arms at the wrist. We will first image able-bodied volunteers, and then repeat the process in two individuals with cervical spinal cord injury.

1.a.ii. PASSIVE AND ACTIVE MOMENTS

Abstract

During the past quarter, we made preliminary measurements to assess the effect of the positions of each finger on the passive properties of adjacent fingers. We found that the position of the metacarpal phalangeal (MP) joint of the long finger had a very strong influence on the passive properties recorded for the MP joint of the index finger. The effect was much larger than anticipated and will be studied further. From the preliminary data, it appears that the influence is due to the skin between the two joints, rather than to tendons or other tissues.

Purpose

The purpose of this project is to characterize the passive properties of normal and paralyzed hands. This information will be used to determine methods of improving hand grasp and hand posture in FES systems.

Report of Progress

During this quarter we made preliminary measurements of the influence of each finger on the passive properties of adjacent digits. As described in the previous progress report, we are working on a device design that has the potential of measuring the passive properties on multiple joints and multiple fingers at the same time. In order to develop some of the design criteria, it was important to know how the positioning of one finger might affect the passive properties measured from an adjacent finger. An existing apparatus was used to make these measurements.

Methods

A subject with no prior history of joint problems was studied (age = 37). Both the index and long fingers of the left hand were splinted so that both the proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints were fixed at 0 degrees extension. The splint attached to the index finger was connected to the passive moment apparatus that has been described previously (QPR#1). This apparatus moves the finger back and forth through its entire range of motion while measuring both joint angle and joint moment. The long finger was strapped down using Velcro straps at various angles of the MP joint.

The long finger MP joint was initially fixed at 5° flexion and the passive torque-angle curve of the index MP joint was recorded using previously established protocols (QPR#2). The long finger MP joint was then positioned at -10, -30°, 65 and 90° and the index MP passive torque-angle curve recorded at each position. At the end of the experiment, the long MP joint was returned to 5° flexion and the measurement repeated.

Results

The passive properties of the index MP joint were strongly influenced by the position of the long MP joint, as shown in Figure 1.a.ii.1. The index MP resting angle changed by nearly 60° for a 120° change in long MP angle. The total range of motion was slightly reduced during extreme flexion or extension of the long MP joint. Figure 1.a.ii.2 shows that there were also changes in the stiffness in both extension and flexion.

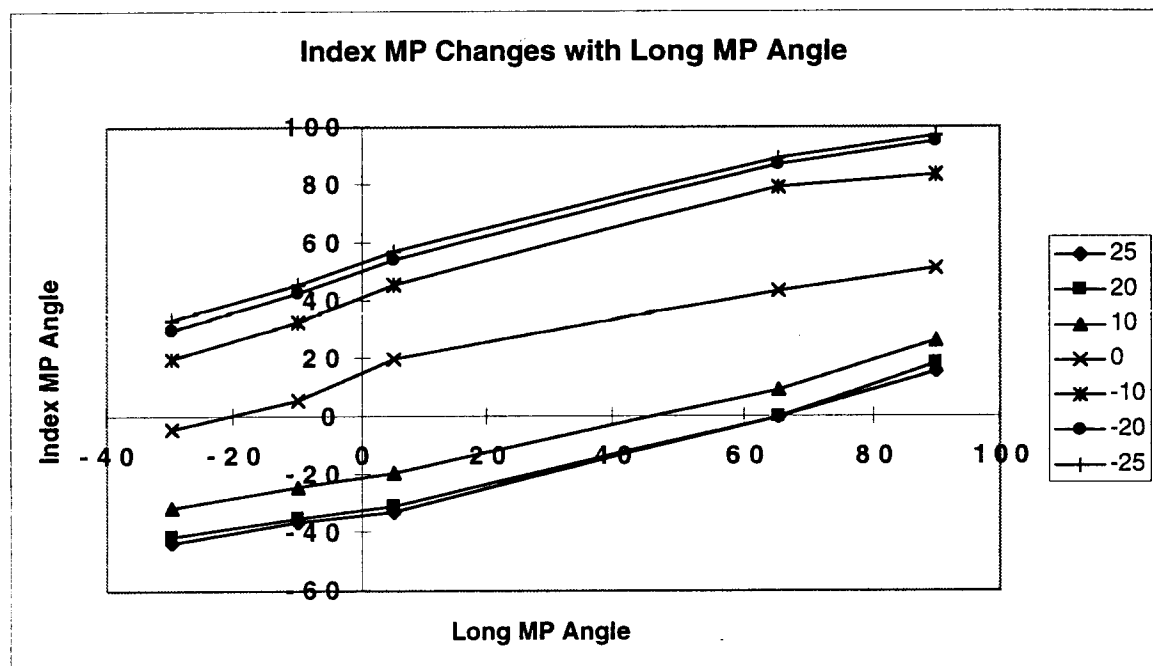


Figure 1.a.ii.1 Index MP range of motion as a function of the MP angle of the long finger. Each line represents the index MP angle at the applied moment indicated in the key. The numbers shown in the key are joint moments in N-cm. The line showing 0 N-cm indicates the rest angle.

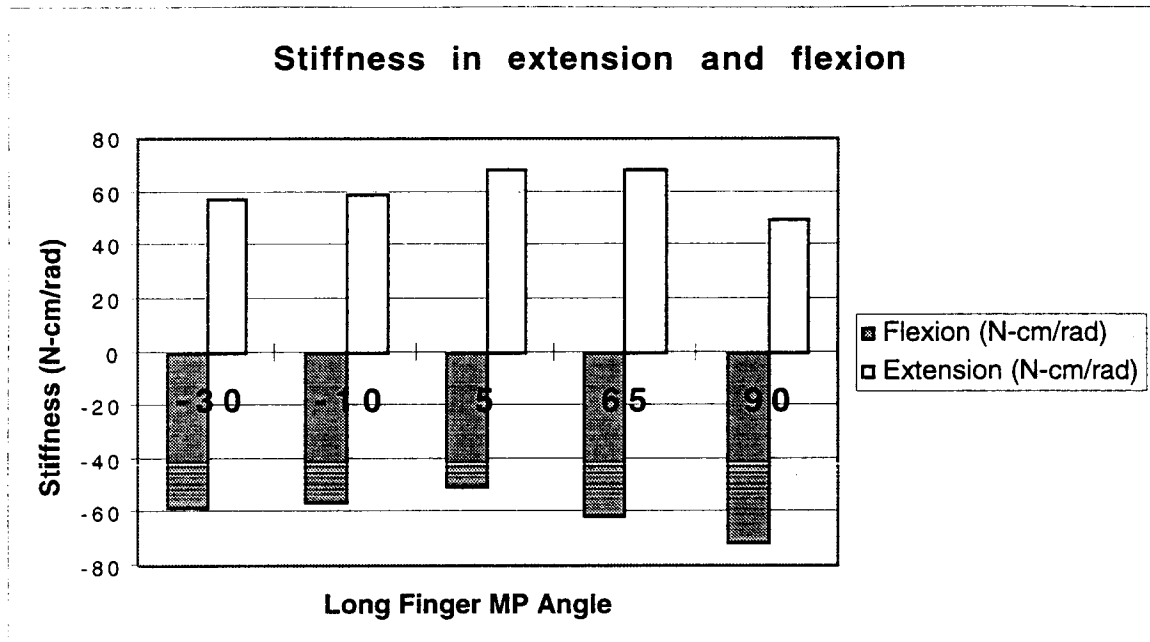


Figure 1.a.ii.2. Stiffness of the index finger between 10 and 20 N-cm of applied moment. Stiffness values are in N-cm/rad. The horizontal axis indicates the angle of the MP joint of the long finger.

Discussion

We anticipated that the position of the long finger would have an effect on the passive properties of the index finger. However, we expected to find that the effect was significant only when the long finger was positioned at extreme flexion or extension. Instead, the results appear to indicate that the effect is significant (and possibly linear) over the whole range of long finger positions. In fact, the change in resting position is nearly as large for long finger movement as it is for the same range of wrist movement (see QPR#4). As far as we know, this effect has not been reported previously. In textbooks describing the measurement of passive range of motion, the position of the adjacent digits is not specified. During the next quarter we will repeat our literature search to see if this effect has been described elsewhere.

Adjacent digits can influence one another either directly through the skin or through linkages between the tendons and muscles (especially in the extensor digitorum communis). The effect that we have measured appears to be due directly to stretching the skin between the two digits because it affects both extension and flexion range equally. In order to verify this, we will repeat these measurements at different wrist angles and with the PIP joint fully flexed. If the effect is purely due to the skin, changing the PIP joint of the long finger should make very little change in the measured properties.

We will continue to evaluate the effect of adjacent digits, including the effect of allowing all the digits to move freely with the joint being measured. Although the results are preliminary, they clearly indicate that it is important to specify the position of all of the digits when passive properties are measured.

Plans for Next Quarter

During the next quarter, we will perform additional experiments to examine the effect of adjacent digit position on passive properties. We will also continue the analysis of the splintless passive moment device described in the previous progress report.

1. b. BIOMECHANICAL MODELING: ANALYSIS AND IMPROVEMENT OF GRASP OUTPUT

Abstract

There was no progress on this aspect of the project this quarter.

Objective

The purpose of this project is to use the biomechanical model and the parameters measured for individual neuroprosthesis users to analyze and refine their neuroprosthetic grasp patterns.

2. CONTROL OF UPPER EXTREMITY FUNCTION

Our goal in the five projects in this section is to either assess the utility of or test the feasibility of enhancements to the control strategies and algorithms used presently in the CWRU hand neuroprosthesis. Specifically, we will: (1) determine whether a portable system providing sensory feedback and closed-loop control, albeit with awkward sensors, is viable and beneficial outside of the laboratory, (2) determine whether sensory feedback of grasp force or finger span benefits performance in the presence of natural visual cues, (of particular interest will be the ability of subjects to control their grasp output in the presence of trial-to-trial variations normally associated with grasping objects, and in the presence of longer-term variations such as fatigue), (3) demonstrate the viability and utility of improved command-control algorithms designed to take advantage of forthcoming availability of afferent, cortical or electromyographic signals, (4) demonstrate the feasibility of bimanual neuroprostheses, and (5) integrate the control of wrist position with hand grasp.

2. a. HOME EVALUATION OF CLOSED-LOOP CONTROL AND SENSORY FEEDBACK

Abstract

The purpose of this project is to deploy an existing portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. We have developed a working prototype of a stand-alone, analog, single channel stimulator for implementing grasp-force feedback, and have also contacted an external vendor to determine the feasibility of modifying a digital, micro-controller based stimulator for the same purpose.

Purpose

The purpose of this project is to deploy a portable hand grasp neuroprosthesis capable of providing closed-loop control and sensory feedback outside of the laboratory. Our goal is to evaluate whether the additional functions provided by this system benefit hand grasp outside of the laboratory.

Report of Progress

We are pursuing the production of a portable, stand-alone (independent of the neuroprosthesis), single-channel stimulator for providing grasp force feedback along two paths. First, we have contacted Axon Engineering (Cleveland OH) to determine whether their existing design for a portable, battery-

powered, 2-channel muscle stimulator can be modified to support grasp force feedback as implemented in our existing portable closed-loop system (with the exception that single pulses are used in lieu of multiple pulse bursts to reduce circuit complexity). The Axon stimulator is based on a 68HC11 micro-controller, and the proposed modifications will allow for a force signal measured from an FSR (as described previously) to be converted to an appropriately-sized output pulse via a look-up table. The look-up table as well as the pulse timing parameters would be down-loaded to the stimulator using a standard serial interface. At present, we are waiting for a cost estimate for producing 2-5 units.

Our second approach is to simplify the overall stimulator design using analog circuitry. We have completed a preliminary design, described below, as well as a working bench prototype. The basic function of the circuit is to modulate the amplitude of a train of pulses with a fixed frequency (e.g. 20 Hz) and pulse width (e.g. 150 μ s) according to the power law transformation described in previous reports, i.e.:

$$i/i_{\min} = \left(F/F_{\min} \right)^m \quad (2.a.1)$$

where i is the pulse current, F is the grasp force, i_{\min} is the threshold current, F_{\min} is the minimum force that produces a perceptible stimulus, and m is given by:

$$m = \frac{\log(i_{\max}/i_{\min})}{\log(F_{\max}/F_{\min})} \quad (2.a.2)$$

with i_{\max} as the maximum acceptable current and F_{\max} as the maximum achievable force. We perform the power law transformation using the AD538 real-time analog computational unit from Analog Devices, which, given three current inputs, executes the first-quadrant function:

$$i_o = i_y \left(\frac{i_z}{i_x} \right)^E \quad (2.a.3)$$

We configure the circuit as shown in Fig. 2.a.1., where the FSR resistance R_f is used to control the current i_z , a fixed resistor R_{\min} programs i_x and is equal to the resistance of the FSR measured at F_{\min} , and a pair of resistors R_b and R_c adjust the exponent E . Incorporating the bases of the two transistors Q1 and Q2 in the resistor divider formed by R1 and R2 assures that:

$$V_x = V_z = V_b + 0.6 = \frac{V_s R_2}{R_1 + R_2} + 0.6. \quad (2.a.4)$$

Therefore, the ratio of the input currents is given by:

$$i_z/i_x = \frac{(V_s - V_z)/R_{\min}}{(V_s - V_x)/R_F} = R_F/R_{\min} \quad (2.a.5)$$

independent of the exact values of the supply voltage or the divider resistors. The FSR resistance can also be approximated well by a power function so that:

$$R_F = \hat{R} \left(F / \hat{F} \right)^{-b} \quad (2.a.6)$$

where \hat{F} is a unit of force (e.g. 1 N) and \hat{R} is the FSR resistance at this force. Solving this expression for the force F and substituting the result into Eq. 2.a.1 yields an expression for the desired current in terms of the FSR resistance:

$$i = i_{\min} \left[\frac{R/\hat{R}}{\left(F_{\min}/\hat{F} \right)^{-b}} \right]^{-m/b} \quad (2.a.7)$$

which simplifies to:

$$i = i_{\min} \left(R_F / R_{\min} \right)^{-m/b} \quad (2.a.8)$$

provided that we choose:

$$R_{\min} = R_F(F_{\min}) = \hat{R} \left(F_{\min} / \hat{F} \right)^{-b} \quad (2.a.9)$$

which can be assured by direct measurement given a pre-selected value of F_{\min} and a particular FSR. We can perform the transformation specified by Eq. 2.a.1 for any subject on any day by allowing the subject (or the attendant) to adjust $i_Y = i_{\min}$ and to adjust the resistors R_B and R_C such that:

$$E = \frac{R_C}{R_B + R_C} = -\frac{m}{b}. \quad (2.a.10)$$

The AD538 circuit was built and tested with $i_Y = 2.0$ mA, $R_B = 68$ k Ω , $R_C = 101$ k Ω ; and with $R_{\min} = 46.4$ k Ω commensurate with the FSR used on the balsa wood compliant object described previously. The maximum current and force were chosen to be 7.3 mA and 10 N, respectively, and the minimum force

was chosen to be 0.5 N. Substituting these values into the equations above predicts a transfer function of:

$$i/i_{\min} = \left(R_F/R_{\min} \right)^{-0.599} \quad (2.a.11)$$

which agrees well with the actual transfer function (derived using DC currents rather than pulses) shown in Fig. 2.a.2:

$$i/i_{\min} = 1.003 \left(R_F/R_{\min} \right)^{-0.580} \quad (2.a.12)$$

The input i_γ to the computational unit is derived from a CMOS monostable multivibrator (CD4538) configured as shown in Fig. 2.a.3. to produce pulses with a width of roughly 150 μ s and a frequency of 20 Hz. The output of the monostable is scaled to produce the desired current i_{\min} (via the voltage input V_γ) via a potentiometer.

Last, the output stage for this prototype is an improved Howland current pump in a bridge configuration, as shown in Fig. 2.a.4, and described in detail in Poletto & Van Doren (1998).

The complete circuit was able to produce perceptible pulses that changed with the force applied to the FSR, though no detailed psychophysical evaluations were performed.

Plans for Next Quarter

The prototype circuit will be evaluated carefully for delivery of rectangular pulses into realistic impedances, and be modified as necessary. A hard-wired, battery powered version will be constructed (using a micropower DC-DC converter) and will be field tested.

Reference:

Reference

POLETTI, C. J. and VAN DOREN, C. L. (1998). A high-voltage constant current stimulator for electrocutaneous stimulation. Accepted for publication in IEEE Trans. Biomed. Eng.

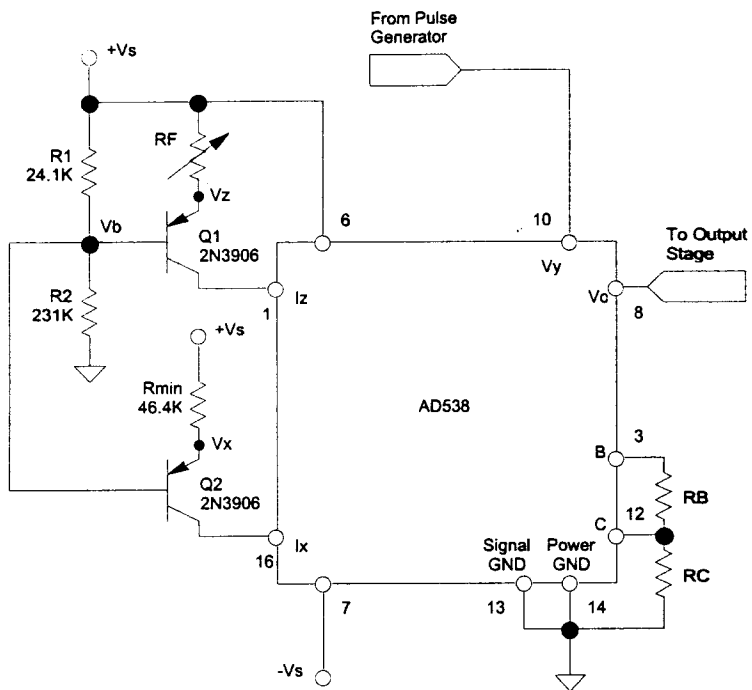


Fig. 2.a.1. Schematic of analog computational unit.

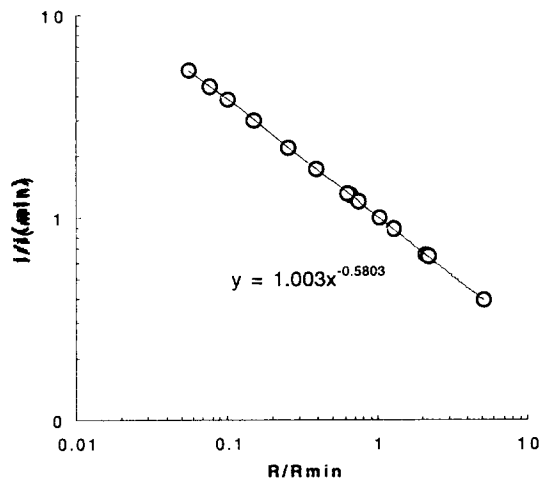


Fig. 2.a.2. Transfer function of prototype force-to-current conversion circuit shown in Fig. 2.a.1, with the component values listed in the text.

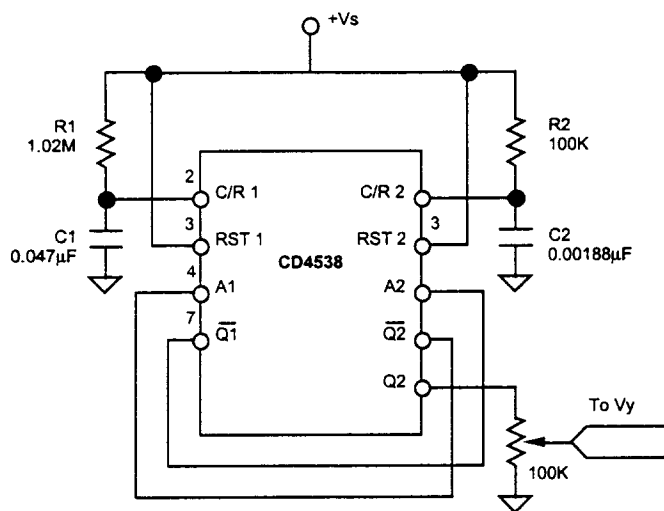


Fig. 2.a.3. Schematic of pulse generator.

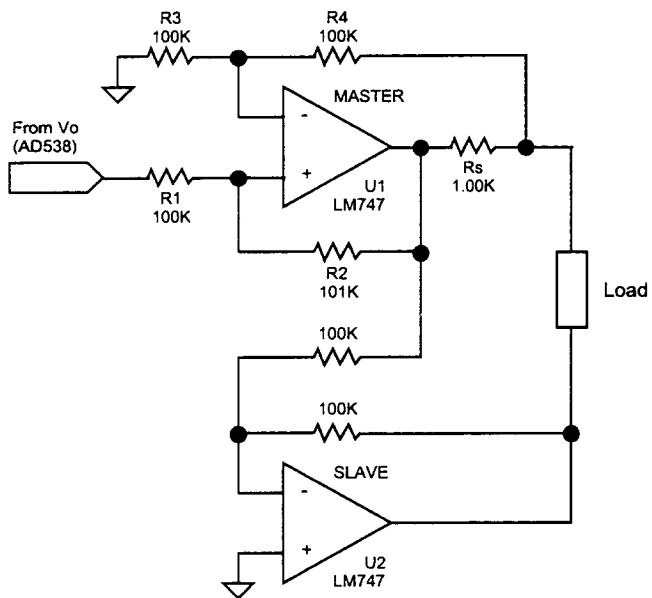


Fig. 2.a.4. Schematic of Howland current pump output stage.

2. b. INNOVATIVE METHODS OF CONTROL AND SENSORY FEEDBACK

2. b. i. ASSESSMENT OF SENSORY FEEDBACK IN THE PRESENCE OF VISION

Abstract

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance in the presence of such visual information. In this quarter, we have developed a command generation strategy by approximating the neuroprosthesis-force transducer system as a piecewise linear system. The necessary software for evaluation of subject performance under different feedback conditions was also developed and tested.

Purpose

The purpose of this project is to develop a method for including realistic visual information while presenting other feedback information simultaneously, and to assess the impact of feedback on grasp performance. Vision may supply enough sensory information to obviate the need for supplemental proprioceptive information via electrocutaneous stimulation. Therefore, it is essential to quantify the relative contributions of both sources of information.

Report of Progress

As discussed in previous reports, realistic visual feedback via digitized video requires a smooth playback of the pre-recorded video clips. Smoothness requires that the neuroprosthesis output change at a constant rate to ensure that there are no abrupt changes between video frames. It is necessary then to modulate the rate change of the input command in order to accommodate for irregularities in the recruitment curves of neuroprosthesis users.

In the previous report, we proposed modeling the complete neuroprosthesis-instrumented object system as a lumped single-input (command), single-output (force) system using a Hammerstein structure which included a static non-linearity (recruitment function) followed by a linear dynamic system (including a delay) [1]. Data for the model was collected from a single neuroprosthesis user. Band-limited noise (5th order, 10 Hz cutoff) was input as the command signal to the neuroprosthesis, and the resulting forces were measured. The system model was then identified using MatLab routines developed in-house. However during verification of the resulting model, the variance accounted for by the model was not sufficient to warrant its use in our control scheme. Several reasons could explain this failure. First, a Hammerstein system may not be appropriate since a Hammerstein model has each of its Wiener kernels identically zero *except* along the main diagonal. Therefore, a nonlinear system whose Wiener kernels are each zero along the main diagonal cannot be approximated by a Hammerstein system or sum thereof. Second, it may be possible that our system cannot be suitably modeled by a Hammerstein/Weiner structure and a different approach such as an LNL structure, kernel, or non-parametric approach may be necessary for modeling the system [1]. Given the time frame for this project further investigation of the problem was not feasible. Our solution to the problem was to approximate the input-output recruitment curve for each individual neuroprosthesis user as a series of straight line segments, with the middle line segments having a larger slope than the end ones (i.e., a piecewise linear system) [2]. The appropriate Command input to the system for a smooth video clip could then be obtained by a simple inversion of this linear system. This method has been tested on a neuroprosthesis user to our satisfaction. Also during this quarter we had to abandon our previous Feedback Evaluation experimental setup due to the failure of a critical component. This necessitated the re-development of the software for performing the evaluations. We have completed and tested the

evaluation software. The main obstacle to the development of this software lay in real time plotting of force trajectories generated by a subject. However this problem has been overcome and the software is now in the form of a free standing application capable of running on any computer hosting a digital I/O card. Some of the salient features of the software are discussed below.

The software interface is identical to the one previously used in our preliminary evaluation trials. The software is capable of reading in a force lookup table specified by the user, and also saving the results of each trial to a file specified by the user. The software allows the user to change the trial time or difficulty level by adjusting the target force window or acquire-hold ratio at any time in between trials. During the course of each trial the software reads in a voltage corresponding to the force being generated by the subject controlling the simulated neuroprosthesis and uses the force lookup table to plot the force in real time display. The software records and provides the user with feedback on the start/end of the trial and success/failure of the subject in each of the acquire, hold, and overall trial stages. The subjects receive audio feedback at the start and end of each trial and in between each trial when the required grasp mode switches from acquire to hold phase.

Plans for Next Quarter

We will be running preliminary evaluation trials shortly using test video clips obtained using our linear approximation Command generation. We will then be scheduling neuroprosthesis users for data collection sessions. The emphasis in the next quarter will be on assembling a large number of video clips for use both in the evaluation of sensory feedback and for assessing control methods.

References

- [1] Durfee WK, MacLean KE (1989) Methods for estimating isometric recruitment curves of electrically stimulated muscle. IEEE Trans BME 36(7)::654-667
- [2] Hines AE, Owens NE, Crago PE (1992) Assessment of input-output properties and control of neuroprosthetic hand grasp. IEEE Trans BME 39: 610-623.

2. b. ii. INNOVATIVE METHODS OF COMMAND CONTROL

Abstract

Due to the technical problems with the computer-based video simulator (see C.2.b.i) there was no progress in this area during the past quarter.

Purpose

The purpose of this project is to improve the function of the upper extremity hand grasp neuroprosthesis by improving user command control. We are specifically interested in designing algorithms that can take advantage of promising developments in (and forthcoming availability of) alternative command signal sources such as EMG, and afferent and cortical recordings. The specific objectives are to identify and evaluate alternative sources of logical command control signals, to develop new hand grasp command control algorithms, to evaluate the performance of new command control sources and algorithms with a computer-based video simulator, and to evaluate neuroprosthesis user performance with the most promising hand grasp controllers and command control sources.

Plans for Next Quarter

In the next quarter we continue command control algorithm evaluation using the video simulator (see C.2.b.i).

2. b. iii. INCREASING WORKSPACE AND REPERTOIRE WITH BIMANUAL HAND GRASP

Abstract

Four able-bodied subjects and one neuroprosthesis user were studied as they started their participation in the cortical control study to identify differences in the active sites and frequencies between able-bodied subjects and persons who have sustained a spinal cord injury. Preliminary results indicate that it will be difficult to operate a neuroprosthesis using the mu rhythm and move the upper extremity at the same time. These subjects also have been unable to attain high levels of mu rhythm control, which has led to the investigation of other sites and frequencies for cortical control. The findings on the mu rhythm, the identification of the new sites, and initial subject results using the new sites are presented.

Purpose

The objective of this study is to extend the functional capabilities of the person who has sustained spinal cord injury and has tetraplegia at the C5 and C6 level by providing the ability to grasp and release with both hands. As an important functional complement, we will also provide improved finger extension in one or both hands by implantation and stimulation of the intrinsic finger muscles. Bimanual grasp is expected to provide these individuals with the ability to perform over a greater working volume, to perform more tasks more efficiently than they can with a single neuroprosthesis, and to perform tasks they cannot do at all unimanually.

Report of progress

In this quarter, data was collected from four able-bodied subjects and one neuroprosthesis user as they entered the cortical control study to identify differences in the motor cortex, both in active sites and active EEG frequencies, which are a result of a spinal cord injury. To achieve this, each subject was asked to participate in three separate protocols, each of which focused upon movement at a different upper extremity joint. In the first protocol, the subject was asked to repetitiously open and close their right hand when a right target appeared on the screen, and to open and close their left hand when a left target appeared. The neuroprosthesis user, since he was unable to generate hand movements, was asked to extend and relax their wrist. Next, subjects were asked to repeat these movements, but with the eyes closed and the investigator informing them of which hand to move. Finally, subjects were asked to again open their eyes and to think about movements of the appropriate hand in conjunction with the target on the screen, but not to actually generate any movements. This whole section was repeated two more times for a total of nine trials for the hand. The other protocols followed the exact series of steps, except that in the second protocol, the subject was asked to flex and extend the wrist, and in the third protocol, the subject was asked to elevate and depress the shoulder.

The results from this study are presented in Tables 2.b.iii.1 - 2.b.iii.7. These tables represent which sites were active and in which frequency bands for each of the five subject. In this table, the mu band was defined as those frequencies between 8 and 13 Hz, while the beta band was defined as those frequencies between 18 and 30 Hz. To determine if a given band was active, an F statistic with a 90%

confidence interval was computed for each 1 Hz frequency bin . If a majority of the frequencies in the bin demonstrated statistical significance, then the band was considered active. For the neuroprosthesis user, we were unable to determine active sites with movement because of the method by which the recording electrodes are attached to the head. The electrodes are imbedded in an elastic cap, and the cap was held in place with two elastic straps which attach to a chest harness. Because the cap was secured to a chest harness, trunk movement caused cap movement, and thus artifacts in the EEG data which made it difficult to interpret. Therefore, data from the neuroprosthesis user on any upper extremity movement, as well as the shoulder movement data from the able bodied subjects, could not be analyzed.

Table 2.b.3.1. Eyes Open Move Hand

	C5	C3	C1	Cz	C2	C4	C6
AB1		β			μ		
AB2				μ	μ	$\mu\beta$	
AB3	β		β	β	β	$\mu\beta$	β
AB4	β	β	β		β	β	β
NP1	*	*	*	*	*	*	*

Table 2.b.3.2. Eyes Open Move Elbow

	C5	C3	C1	Cz	C2	C4	C6
AB1		$\mu\beta$		β	μ		
AB2	μ	μ	$\mu\beta$	$\mu\beta$	$\mu\beta$	μ	μ
AB3	$\mu\beta$		μ			β	β
AB4	μ		β	β	$\mu\beta$	β	β
NP1	*	*	*	*	*	*	*

Table 2.b.3.3. Eyes Closed Move Hand

	C5	C3	C1	Cz	C2	C4	C6
AB1	μ	β					μ
AB2	μ		β	β	$\mu\beta$		
AB3			—	—		—	
AB4	—		—	—	—	—	
NP1	—	—	—	—	—	—	—

Table 2.b.3.4. Eyes Closed Move Elbow

	C5	C3	C1	Cz	C2	C4	C6
AB1		μ	β	β	β	β	
AB2							
AB3	—		—			—	
AB4	—		—		—		—
NP1	—	—	—	—	—	—	—

Table 2.b.3.5. Eyes Open Think Hand

	C5	C3	C1	Cz	C2	C4	C6
AB1	—	—	—	—			
AB2					—	—	—
AB3		—			—		
AB4				—		—	
NP1				—			

Table 2.b.3.6. Eyes Open Think Elbow

	C5	C3	C1	Cz	C2	C4	C6
AB1		—				—	—
AB2					—		—
AB3	—	—		—	—	—	—
AB4	—		—				
NP1							

Table 2.b.3.7. Eyes Open Think Shoulder

	C5	C3	C1	Cz	C2	C4	C6
AB1	—	—	—	—	—	—	—
AB2			—			—	
AB3	—	—	—	—	—	—	—
AB4	—		—	—	—	—	
NP1	—	—		—	—	—	—

*Movement artifact in the data

—Did not conduct the experiment

The results of this study indicated that there are no specific sites active for hand, elbow, or shoulder movement. However, this would be misleading since there are other factors that may have contributed to the 'random' activation that was displayed in the data. The first was that although each subject was instructed to remain completely relaxed so as to isolate the specific muscle group being examined, many other muscles were also activated. Even a very simple movement about one joint requires simultaneous coordination of many muscles (Humphrey 1986). For example, when the actual movement of grasp and release of the hand occurs, both the flexor carpi (FCR) and extensor carpi radialis (ECR) contribute to wrist fixation, while the deltoid contributes to stabilization of the shoulder. During extension and flexion of the elbow, again both the FCR and ECR contribute to wrist fixation, while the deltoid stabilizes the shoulder. Finally, during elevation and depression of the shoulder, both the biceps brachii and the triceps are active to contribute to elbow fixation. There also existed the possibility that the leg and/or trunk muscles were also active since subjects were seated in an upright position, which could account for Cz, C1, and C2 activation. Finally, in the trials where the subjects eyes were open, as they watched the target on the screen, their eye movement was perfectly correlated with limb movement and could account for any C5 and C6 activation.

The use of the electrodes imbedded in an elastic cap also introduced substantial problems. First, the "10-20 system" of electrode placement, which the cap utilizes, uses scalp landmarks to place electrodes. The basic assumption is that these scalp locations consistently correlate with underlying cerebral structures, which may not necessarily always be true. Binnie, et al. (1982) found that the cerebral quadrants bounded by the nasion, anion and preauricular points varied in size by more than 10% in the majority of normal subjects. Homan, et al. (1987) found that 17.7% of the electrode positions deviated substantially from the predominant scalp location. They also found substantial variability of cortical marker locations among individuals. This variability, combined with the fact that the same size cap was used for all subjects despite differences in scalp size and geometry and that only the nasion to anion distance was measured, can lead to major complications in locating a specific site. Finally, and perhaps most importantly, the distance between electrode locations on the cap is much too large to differentiate between regions for hand, elbow, and shoulder movement. Cohen et al. (1991) found only a 2 centimeter range (5-7 cm down the motor cortex from Cz) in scalp location for flexor carpi radialis, biceps, and deltoid activation along with considerable overlap. The electrodes on the cap cover a circular region of approximately 2 to 2.5 centimeters in diameter. This places the hand, elbow, and shoulder locations all in the C3 and C4 electrode region.

The data collected from this study provided us with two very important findings. The first of these was that it is difficult, given the equipment and methodology, to determine specific areas which are active with hand, elbow and shoulder movement. Secondly, and more importantly, it has been concluded that it would be difficult to control the mu rhythm and perform movements of the upper extremity at the same time. This has been verified in a separate study where one of the able-bodied subjects with a high level of mu rhythm control (greater than 90% accuracy rate in hitting targets) attempted to lift and move a 0.5 kg weight with both the right and left hands. The weight had to be lifted across a line drawn on a table in front of the subject that corresponded with the mid-line of the body. This was felt to be representative of a neuroprosthesis user moving an object with their system. The spectral plots for the C3 and C4 sites with normal cursor control, right hand movement, and left hand movement are shown in Figures 2.b.3.1 and 2.b.3.2. From these figures, it can be seen that the subjects control of the mu rhythm degenerated with upper extremity movement to such a point that no mu rhythm separation could be determined.

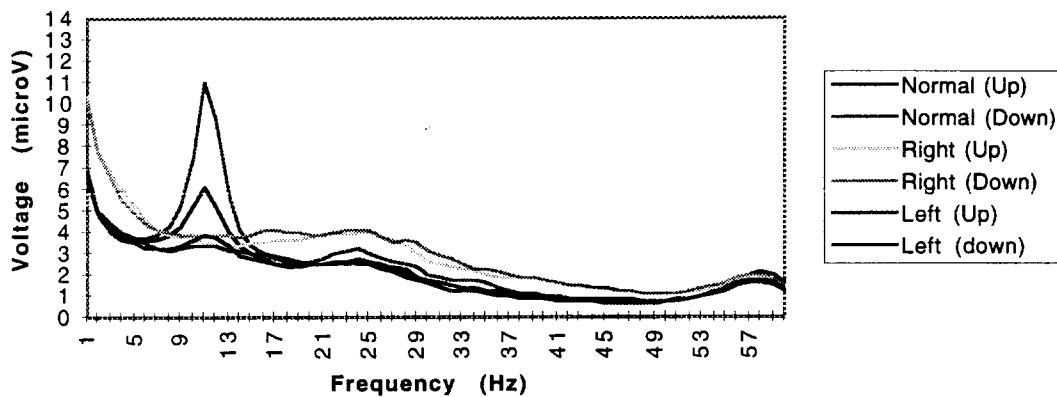


Figure 2.b.3.1 : Spectral Plot of the mu rhythm voltages at the C3 site with no movement (normal), right hand movement (right) and left hand movement (left) at the C3 site.

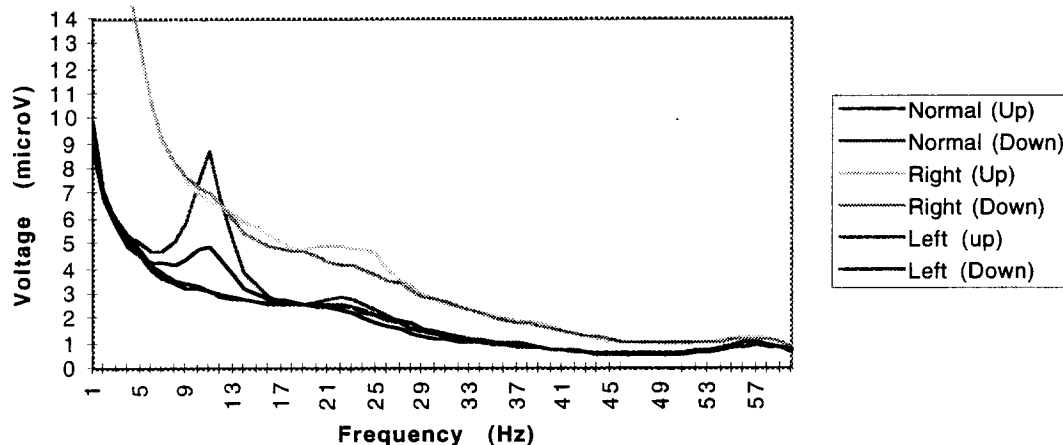


Figure 2.b.3.2 : Spectral Plot of the mu rhythm voltages at the C4 site with no movement (normal), right hand movement (right), and left hand movement (left) at the C4 site.

An investigation was begun to study alternative sites and frequencies for cortical control in order to overcome the problems identified above. Literature on biofeedback and on the operation of similar systems to the BCI indicated that there are several options for the use of cursor control: 1) alpha rhythm, 2) slow cortical potentials, 3) evoked potentials, and 4) beta rhythm.

The alpha rhythm is defined as the 8-13 Hz component of the EEG recorded over the posterior regions of the head, generally with higher voltages over the occipital areas (Sites O1, O2, P3 and P4). The frequency is best observed in the EEG signal when the eyes are closed and the person is relaxed both physically and mentally (Kooi, et al. 1978; Niedermeyer, et al. 1982). This frequency can be blocked or attenuated by attention, especially visual, and mental effort. Numerous studies have demonstrated that subjects can be trained to voluntarily control the amplitude of the alpha rhythm both with visual (Tyson 1982; Jenkins and Moore 1985) and auditory feedback (Fox 1979; Drennen and O'Reilly 1986). These

studies also indicate that the alpha rhythm can be recorded from either the occipital (Davidson 1988; Fehmi and Sundor 1989; Saxby and Peniston 1995), temporal (Suter, et al. 1981; Kotchoubey, et al. 1996) or frontal (Rosenfeld, et al. 1995; Hardman, et al. 1997) areas of the brain. Since these areas of the brain are not responsible for voluntary or involuntary movement, the alpha rhythm becomes an ideal candidate for a substitute to the mu rhythm recording.

Slow cortical potentials are a slow developing DC offset in the EEG signal which subjects can be trained to control in a similar manner as the BCI system (Elbert, et al. 1980; Lutzenberger, et al. 1980; Birbaumer, et al. 1988; Roberts, et al. 1989). However, these offset shifts are slow to develop, in some cases requiring 500 ms to begin changing towards a positive or negative voltage. Since this delay of 500 ms is unacceptable for the control of a neuroprosthesis (Graupe, et al. 1984), this option will not be investigated.

Evoked potentials, or the P300, is a distinct voltage potential change in the EEG signal that is the result of the subject experiencing an expected stimulus (either visual, auditory, or tactile). While there have been numerous studies on subjects training to control the evoked potential and using this to operate various systems (Ruchkin, et al. 1981; Petten and Kutas 1988; Sommer and Schweinberger 1992), its use for the neuroprosthesis is limited since it is an evoked response to a stimulus, and not something which can be voluntarily controlled. Also, the development of this potential requires a minimum of 300 ms (thus the name P300), which may be too long of a delay for satisfactory neuroprosthetic operation.

The beta rhythm is any frequency in the band above 13 Hz but below 70 Hz. It is recorded over the frontal and central regions (Sites F3, F4, C3 and C4) , with that recorded over the central regions being closely linked to the mu rhythm (Pfurtscheller 1981; Pfurtscheller, et al. 1997). The frontal beta does not have any physiological correlates, and is fairly common, occurring in 80 to 95% of the population (Niedermeyer, et al. 1982). Since this rhythm has no physiological correlates, and can be controlled by mental effort , it is also an ideal candidate for mu rhythm replacement.

Currently, 3 of the 5 subjects enrolled in the study (2 able bodied subjects and 1 neuroprosthesis user) have begun training with these other sites. Although only 1 - 2 sessions have been completed with these subjects at this time, the results were very encouraging. Subject AB-1, when attempting mu rhythm control, also demonstrated a strong alpha rhythm response at the O1 site. This subject has started retraining at this site with a 89.9 % accuracy rate (1 session). Subject AB-2, on the other hand, demonstrated a strong beta response at the FP2 and FP1 sites, and when switched to these sites, exhibited a 92% accuracy rate after 2 sessions. Subject NP-1, on his initial trial, attempted to control the cursor using the alpha rhythm at the O1 site. Overall accuracy after this session was only 50.4%, which led to further investigation of other sites in this subject. It was found that this subject, like AB-2, has a strong beta signal in the frontal sites (F3 and AF7), and when trained using these sites, exhibited a 96.7 % accuracy rate.

Plans for Next Quarter

During the next quarter, all subjects will be switched over to the alternate sites and frequencies, and will be trained in the control of these signals to operate cursor control. As the subjects achieve a greater than 90% accuracy rate and have demonstrated that they can repeatedly accomplish this, they will be asked to move the cursor on the computer screen while performing upper extremity movements to verify that there is no interaction between movement and the signals at these sites. During this time, the interface between the BCI and neuroprosthesis will be developed so that testing on the operation of the hand grasp system using cortical signals can begin.

References

- Binnie, C.D., Dekker, E., Smith, A. and Van der Linden, G. (1982). "Practical considerations in the positioning of EEG electrodes". *Electroencephalography and clinical Neurophysiology* 53: 453-458.
- Birbaumer, N., P. Lang, et al. (1988). "Slow brain potentials, imagery and hemispheric differences." *Int J Neurosci* 39((1-2)): 101-116.
- Cohen, L.G., Bandinelli, S., Topka, H.R., Fuhr, P., Roth, B.J., Hallett, M. (1991). "Topographic Maps of Human Motor Cortex in Normal and Pathological Conditions: Mirror Movements, Amputations and Spinal Cord Injuries" *Magnetic Motor Stimulation: Basic Principles and Clinical Experience* 43: 36-50.
- Davidson, R. (1988). "EEG Measures of Cerebral Asymmetry: Conceptual and Methodological Issues." *Intern J Neuroscience* 39: 71-89.
- Drennen, W. and B. O'Reilly (1986). "Alpha enhancement: A comparison study of biofeedback, open focus training, and control procedures." *Perceptual and Motor Skills* 62: 467-474.
- Elbert, T., B. Rockstroh, et al. (1980). "Biofeedback of slow cortical potentials. I." *Electroencephalogr Clin Neurophysiol* 48(3): 293-301.
- Fehmi, L. and A. Sundor (1989). "The effects of electrode placement upon EEG biofeedback training: the monopolar-bipolar controversy." *Int J Psychosom* 36((1-4)): 23-33.
- Fox, C. (1979). "Feedback control of hemispheric EEG alpha." *Percept Mot Skills* 48(1): 147-155.
- Graupe, D., K. Kohn, et al. (1984). "Electromyographic control of functional electrical stimulation in selected patients." *Orthopaedics* 7(7): 1134-1138.
- Hardman, E., J. Gruzelier, et al. (1997). "Frontal interhemispheric asymmetry: self regulation and individual differences in humans." *Neurosci Lett* 221(2-3): 117-120.
- Homan, R.W., Herman, J., and Purdy, P. (1987) "Cerebral location of international 10-20 system electrode placement". *Electroencephalography and clinical Neurophysiology* 66: 376-382.
- Humphrey, D.R. (1986). "Representation of movements and muscles within the primate precentral motor cortex: historical and current perspectives". *Federation Proc.* 45: 2687-2699.
- Jenkins, P. and W. Moore (1985). "The effects of visual feedback on hemispheric alpha asymmetries and reported processing strategies: a single-subject experimental design." *Brain and Cognition* 4: 47-58.
- Kooi, K., R. Tucker, et al. (1978). *Fundamentals of Electroencephalography*. New York, Harper & Row.
- Kotchoubey, B., H. Schleichert, et al. (1996). "Self-regulation of interhemispheric asymmetry in humans." *Neurosci Lett* 215(2): 91-94.
- Lutzenberger, W., T. Elbert, et al. (1980). "Biofeedback of slow cortical potentials. II. Analysis of single event-related slow potentials by time series analysis." *Electroencephalogr Clin Neurophysiol* 48(3): 302-311.
- Niedermeyer, E. and F. L. d. Silva (1982). *Electroencephalography: Basic Principles, Clinical Applications and Related Fields*. Baltimore, Urban & Schwarzenberg.
- Petten, C. V. and M. Kutas (1988). "The use of event-related potentials in the study of brain asymmetries." *Intern J Neuroscience* 39: 91-99.
- Pfurtscheller, G. (1981). "Central Beta Rhythm during Sensorimotor Activities in Man." *Electroenceph Clin Neurophysiol* 51: 253-264.
- Pfurtscheller, G., A. Stancak, et al. (1997). "On the existence of different types of central beta rhythms below 30 Hz." *Electroenceph Clin Neurophysiol* 102: 316-325.
- Roberts, L., N. Birbaumer, et al. (1989). "Self-report during feedback regulation of slow cortical potentials." *Psychophysiology* 26(4): 392-403.
- Rosenfeld, J., G. Cha, et al. (1995). "Operant (biofeedback) control of left-right frontal alpha power

differences: potential neurotherapy for affective disorders." *Biofeedback Self Regul* 20(3): 241-258.

Ruchkin, D., S. Sutton, et al. (1981). "P300 and feedback provided by absence of the stimulus." *Psychophysiology* 18(3): 271-282.

Saxby, E. and E. Peniston (1995). "Alpha-theta brainwave neurofeedback training: an effective treatment for male and female alcoholics with depressive symptoms." *J Clin Psychol* 51(5): 685-693.

Sommer, W. and S. Schweinberger (1992). "Operant conditioning of P300." *Biol Psychol* 33(1): 37-49.

Suter, S., G. Griffin, et al. (1981). "Biofeedback regulation of temporal EEG alpha asymmetries." *Biofeedback Self Regul* 6(1): 45-56.

Tyson, P. (1982). "The choice of feedback stimulus can determine the success of alpha feedback training." *Psychophysiology* 19(2): 218-230.

2. b. iv CONTROL OF HAND AND WRIST

Abstract

In this quarter, we completed and submitted a manuscript describing our previous work on coordinated feedforward control of wrist angle and hand grasp. We have also initiated a project to examine the biomechanics of the extensor carpi ulnaris to extensor carpi radialis brevis transfer, which is an important component of restoring wrist extension. Toward this goal, we have constructed a wrist moment transducer.

Purpose

The goal of this project is to design control systems to restore independent voluntary control of wrist position and grasp force in C5 and weak C6 tetraplegic individuals. The proposed method of wrist command control is a model of how control might be achieved at other joints in the upper extremity as well. A weak but voluntarily controlled muscle (a wrist extensor in this case) will provide a command signal to control a stimulated paralyzed synergist, thus effectively amplifying the joint torque generated by the voluntarily controlled muscle. We will design control systems to compensate for interactions between wrist and hand control. These are important control issues for restoring proximal function, where there are interactions between stimulated and voluntarily controlled muscles, and multiple joints must be controlled with multijoint muscles.

Report of progress

The manuscript describing our simulation and experimental demonstration of coordinated integrated control of hand grasp and wrist angle was completed and submitted to a journal for review. The authors, title and abstract are as follows.

Authors: Margaret M. Adamczyk and Patrick E. Crago

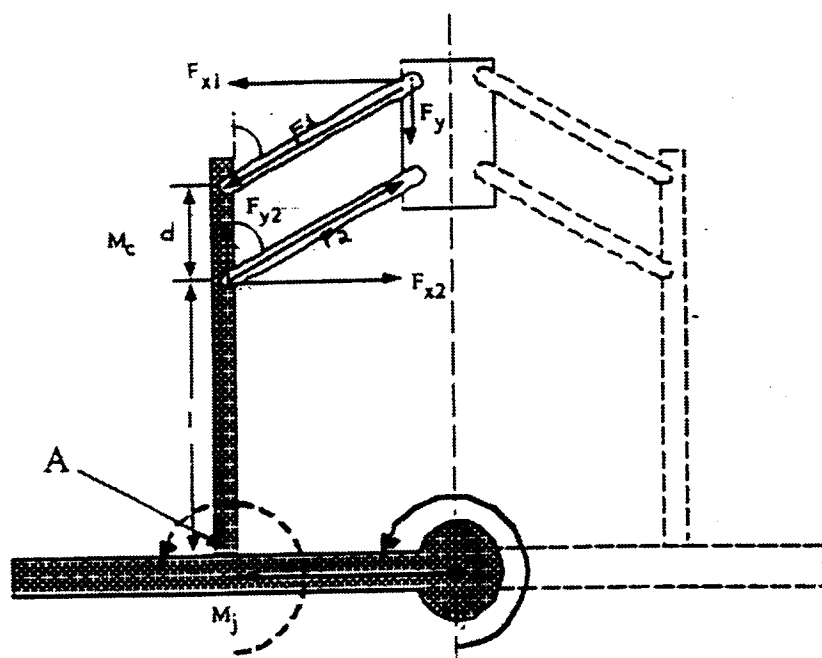
Title: Feedforward neural network coordination of hand grasp and wrist angle in a neuroprosthesis

Abstract: This study addresses a specific example of the general problem of coordinating muscle stimulation in a neuroprosthesis when multiarticular muscles introduce mechanical coupling between joints. In a hand grasp neuroprosthesis, extrinsic hand muscles cross the wrist joint and introduce large wrist flexion moments during grasp. In order to control hand grasp and wrist angle independently, a controller must take the mechanical coupling into account. In simulation, and in a clinical demonstration, we investigated the use of feedforward artificial neural networks to coordinate hand and wrist muscle stimulation. The networks were trained with data that is easily obtained experimentally. Feedforward control showed excellent hand and wrist coordination when the properties of the system were fixed and there were known external loads. However, the low stiffness at the wrist resulted in significant angle errors if muscle gains changed or if moderate external loads were applied. These errors

We have also completed construction of a device for measuring isometric wrist moments. This is a mechanical linkage system to measure the moments at the wrist joint with two degrees of freedom, one in the flexion/extension plane the other in the radial/ulnar plane. Since the moments are measured directly they are more reliable than measuring force and finding the center of rotation, as we have done in the past. The moment measurement is not subject to the position of placement of the instrument. The design is an extension of the principles used previously in instruments to measure finger moments and elbow moments [Kilgore et al. 1998].

For equilibrium the forces F_1 and F_2 in the links must balance. Since frictionless pivots connect the links only the forces are transferred to the uprights. This force will produce a bending moment in the upright that is given by couple $M_c = F_{x1} * d$. The only other moment is joint torque $M_j = M_c$.

additional set of links proximally in the horizontal plane that produce a bending moment in the upright corresponding to the radial/ulnar moments at the wrist.



The instrument has been machined. The strain gauges are being mounted and it will be calibrated before making the moment measurements. The device has a range of operation from 60° flexion to 60° extension, and in the radial/ulnar plane only the physiological range of motion would limit the angle for the subject. It is designed to be sensitive to moments of 0.2N/cm and have a resolution of approximately 0.3N/cm.

We are currently calibrating the device, and after calibration, we will make wrist moment measurements in

patients who are going to receive tendon transfers, and those who have had them previously.

Kilgore, K.L., Lauer, R.J., and Peckham, P.H., A transducer for the measurement of finger joint moments, IEEE Trans. Rehab. Eng., in press